Active humidification for capacitive-resistive ECG-systems

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Abstract
Capacitive ECG measurement systems have been moving in the research spotlight in the past years. Due to their ability to measure ECG signals without direct contact to the body surface, they are well-suited for monitoring applications in personal healthcare scenarios. However, signal quality is insufficient for diagnostic purposes. Main causes are motion artefacts and triboelectricity. In this work, we propose active humidification using water vapour as a measure to improve signal quality. After a brief theoretical analysis and a proof of concept, we present a prototype of a capacitive ECG seat implementing the concept. Using this prototype, we obtained experimental proof that active humidification can in fact improve cECG signal quality.

1 Introduction
In 1967, Richardson first described the derivation of an electrocardiogram (ECG) using insulated, that is capacitive electrodes [1]. Signal acquisition with this electrode type does not require a conductive (galvanic) connection to the patient under examination, since ECG recording is accomplished by means of an electrical field: The patient’s body surface and the electrode form a structure similar to a parallel-plate capacitor, thus by electrostatic induction, potentials present on the body surface can also be measured at the electrode. Using a field based, non-contact measurement principle, no electrodes have to be attached to the skin (which can be irritating) and the ECG can be recorded even through layers of clothing.

Due to this benefits, during the past years this electrode type has come into research spotlight: Capacitive ECG measurement systems have been integrated in several items of daily living, for example a toilet seat [2], a bathtub [3], a car seat [4] and a tablet-PC based system that can record body surface potential maps [5]. Also, a clinical proof of practicability of a cECG device has recently been shown [6].

However, cECG measurement systems require carefully designed ultra-high input impedance amplifier circuits with additional bias current compensation [7]. Recently, a cECG system has been proposed which does not require a Driven Right Leg Electrode [8]. Still, signal quality of cECG systems is inferior to conventional, conductive ECG systems; in particular in environments with a high level of motion between electrode and body surface, such as while driving on a bumpy road [9]. The causes are for one a bad and time variable capacitive coupling between electrode and body surface causing so called motion artefacts [10]. Due to their randomness, motion artefacts are difficult to separate from the cECG signal using conventional techniques such as filtering, but stochastic approaches such as Blind Source Separation (BSS) have shown promising results [11]. In addition to motion artefacts, electrostatic charge generation can occur on all interfaces between electrode, clothing and skin, for example between a wool sweater and the electrode. This triboelectricity can easily reach magnitudes in the kilovolt range [12], masking the weak ECG signal, which is in the millivolt range. Providing a discharge path that can remove triboelectric charges using an actively driven grid (as analysed in [12]) can only remove charges on the surface of the clothing next to the electrode. Charges generated inside the fabric or another clothing layer flow only slowly to the grid as dry fabric has a very high impedance (usually Gigaohm range), so the effect of the grid is limited.

Here, we propose a different approach that can improve the signal quality in cECG systems by addressing both the poor capacitive coupling between electrode and body surface as well as triboelectric charge generation at the interfaces.

2 Methods
2.1 Analysis
Figure 1 shows a simplified equivalent circuit diagram of a cECG measurement setup, including the ECG signal on the body surface (represented by a voltage source), the capacitive interface (represented by a RC network to account for the capacitive coupling \((C_c)\) with a high impedance residual resistive component \((R_c)\)) and an operational amplifier as the first stage of an ECG signal processing chain [13]. Cable capacitance and parallel input capacitance of the amplifier are accounted by a parasitic capacitor \(C_{in}\). \(R_{in}\) represents the amplifier’s input resistance and the parallel bias resistor commonly used for bias current compensation [7]. The complex transfer function of the setup can be calculated according to equation 1 as

\[
H(j\omega) = \frac{R_{in}}{R_{in} + R_c \cdot \frac{1+j\omega R_c C_{in}}{1+j\omega R_c C_c}}.
\]

(1)
Figure 1 Simplified equivalent circuit, adapted from [13].

Evaluating the equation shows that small coupling resistances in combination with very high input amplifier resistances are beneficial as in this case the transfer function becomes real valued and independent of frequency. The measurement setup then turns into a common galvanic ECG measurement in which coupling capacities are neglectable due to the small coupling resistance. Therefore, due to the improved transfer function, a better ECG signal quality can be expected.

In prior experiments using an ECG system integrated into a car seat we observed an improving signal quality after the test person had been sitting on the seat for several minutes. The effect seemed to be better the more the test person was sweating. Based on this empirical finding, we formulated the thesis that humidity provided by the test person while sweating might condensate inside the clothing and lead to a reduced fabric impedance. This will reduce the coupling resistance. The cECG then changes into a capacitive-resistive ECG measurement system. We also expect that a reduced fabric impedance facilitates the discharge of local triboelectric charges using current paths inside the fabric or via the reduced coupling resistance to the electrode. To benefit from the effect of condensing humidity in measurement situations with little or no transpiration, we propose an implementation of a cECG measurement system that can actively dispense small quantities of water vapour to its electrodes and the clothing of a test person. The practicability and effectiveness of such a system was evaluated in this paper.

2.2 Proof of concept

According to our thesis, condensing humidity can reduce fabric impedances. For verification, we exhibited a small piece (length: 152 mm, width: 40 mm) of cotton fabric to water vapour from a bowl of hot water and measured the fabric impedance on a length of 28 mm for 14 minutes using a Agilent E4980A Precision LCR Meter (Agilent Technologies, Santa Clara, USA). Every minute, a measurement was taken. The results are shown in figures 2 and 3. During the experiment, the measured impedance significantly diminished from 760 MΩ $\angle -89^\circ$ to 23 MΩ $\angle 0^\circ$. Hence, the capacitive coupling was transformed to a low impedance resistive coupling, which proves the thesis. After the experiment, the fabric felt moistly, but not wet.

Figure 2 Resistance development during active humidification.

Figure 3 Phase development during active humidification.

2.3 Experimental setup

To investigate the effect of an electrode with active humidification on ECG signal quality, we constructed a prototype. In our experimental setup, we used an ordinary car seat in which two electrodes made of a flexible, conductive fabric were integrated as shown in figure 4.

Behind the electrodes a device that can release humidity as water vapour was attached. The electrode material is water permeable, so humidity can pass through the electrode into the clothing of a user sitting on the car seat as shown in figure 5.

For ECG derivation, we used a common analog ECG interface. The signals from the two textile electrodes are directly fed into the inputs of an INA116 instrumentation amplifier (Texas Instruments Incorporated, Dallas, USA) [7], processed in a signal processing chain (lowpass-, notch-, highpass filter and amplifier) and are afterwards digitised. In order not to diminish the input impedance $R_i$ for best transfer function performance, no bias resistor for bias current compensation was used as proposed by [7]. To reject common mode disturbances, a Driven Right Leg Circuit was used. The digitised signals were digitally filtered (lowpass, notch, highpass) using Matlab (The Mathworks Inc., Natick, USA).
2.4 Experimental trial

In the experiment two healthy male test subjects were enrolled. To create a realistic measurement scenario, both were wearing two layers of own clothing, a cotton T-Shirt and a cotton sweater. For each test subject, we first recorded 5 min of ECG data using the measurement setup with deactivated humidification device to acquire a dry reference measurement. Then, the device was activated and after 5 min given for humidification, we recorded another 5 min sequence. After each test subject, the whole system was dried to remove residual humidity, so a dry reference measurement was possible for the next test subject. During all measurements, we recorded a conventional 3-lead ECG using a Philips MP30 patient monitor and adhesive Ag/AgCl electrodes as reference ECG.

3 Results

Figure 6 shows a 10 s signal sequence recorded using dry electrodes, figure 7 a 10 s signal sequence using humidified electrodes, respectively. In both figures, the upper graph shows the reference measurement, the lower graph the capacitive measurement. Qualitative data analysis reveals that during the dry measurement, the derived cECG signal is very erratic. The signal shows quick changes with high amplitudes that are not aligned with the R-peaks in the reference ECG. Overall, signal quality is very poor and no clear ECG signal can be observed.

In contrast, in the sequence recorded with humidified electrodes, signal levels are smaller, but a typical ECG signal can be observed which is well aligned to the reference measurement. All relevant parts of an ECG cycle (P-wave, QRS-complex, T-wave [14]) are visible. Even so, one larger artefact is present in the signal, which was recorded while the test subject took a deep breath.

For better comparison, the amplitude of the strongest signal part of an ECG, the R-peak was measured and the amplitude of the residual noise floor for both measurement scenarios was estimated. To derive a quantitative measurement for signal quality, the signal to noise ratio (SNR) between
R-peak amplitude and noise floor was calculated according to equation 2:

\[
SNR = \frac{U_{R-peak}}{U_{Noise floor}}
\]  

(2)

The measured amplitudes and calculated SNR are shown in table 1.

<table>
<thead>
<tr>
<th></th>
<th>Dry electrode</th>
<th>Wet electrode</th>
</tr>
</thead>
<tbody>
<tr>
<td>R-Peak [V]</td>
<td>1.2</td>
<td>0.14</td>
</tr>
<tr>
<td>Noise Floor [V]</td>
<td>1.2</td>
<td>0.04</td>
</tr>
<tr>
<td>Derived Q-factor</td>
<td>1</td>
<td>3.5</td>
</tr>
</tbody>
</table>

**Table 1** Derived quality markers

As can be seen, the ratio between R-peak and amplitude and residual noise floor could be improved by a factor of 3.5 using active humidification. Therefore, a qualitative and quantitative improvement was observed.

4 Conclusion and Outlook

In this work, a new concept for cECG signal quality enhancement using active humidification was proposed. To evaluate the concept, a seat integrated cECG measurement system was constructed. For two test subjects, the effectiveness of the concept was successfully demonstrated.

In future work, cECG data from a larger number of test subjects wearing different types of clothing should be collected to gather statistically relevant data.

5 References


